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published in
Ergonomics
2019

DOI (link to publisher)
10.1080/00140139.2019.1602288

document version
Publisher's PDF, also known as Version of record

citation for published version (APA)
Baltrusch, S. J., van Dieën, J. H., Bruijn, S. M., Koopman, A. S., van Bennekom, C. A. M., & Houdijk, H. (2019). The effect of a passive trunk exoskeleton on metabolic costs during lifting and walking. *Ergonomics*, *62*(7), 903-916. https://doi.org/10.1080/00140139.2019.1602288

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ERGONOMICS
2019, VOL. 62, NO. 7, 903–916
https://doi.org/10.1080/00140139.2019.1602288

The effect of a passive trunk exoskeleton on metabolic costs during lifting and walking

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ABSTRACT
The objective of this study was to assess how wearing a passive trunk exoskeleton affects metabolic costs, movement strategy and muscle activation during repetitive lifting and walking. We measured energy expenditure, kinematics and muscle activity in 11 healthy men during 5 min of repetitive lifting and 5 min of walking with and without exoskeleton. Wearing the exoskeleton during lifting, metabolic costs decreased as much as 17%. In conjunction, participants tended to move through a smaller range of motion, reducing mechanical work generation. Walking with the exoskeleton, metabolic costs increased up to 17%. Participants walked somewhat slower with shortened steps while abdominal muscle activity slightly increased when wearing the exoskeleton. Wearing an exoskeleton during lifting decreased metabolic costs and hence may reduce the development of fatigue and low back pain risk. During walking metabolic costs increased, stressing the need for a device that allows disengagement of support depending on activities performed.

Practitioner summary: Physiological strain is an important risk factor for low back pain. We observed that an exoskeleton reduced metabolic costs during lifting, but had an opposite effect while walking. Therefore, exoskeletons may be of benefit for lifting by decreasing physiological strain but should allow disengagement of support when switching between tasks.

Abbreviations: COM: centre of mass; EMG: electromyography; LBP: low back pain; MVC: maximum voluntary isometric contraction; NIOSH: National Institute for Occupational Safety and Health; PLAD: personal lift augmentation device; PWS: preferred walking speed without exoskeleton; PWSX: preferred walking speed with exoskeleton; ROM: range of motion; RER: respiratory exchange ratio; $\dot{V}O_2max$: maximum rate of oxygen consumption

ARTICLE HISTORY
Received 17 April 2018
Accepted 25 March 2019

KEYWORDS
Assistive device; low back pain; oxygen consumption; movement behaviour; EMG

1. Introduction

Most adults (60–80%) experience low-back pain (LBP) at some point in their lifetime (Waddell and Burton 2001). Many suffer from relapses of pain (44–78%) and work absence (26–37%) (Hestbaek, Leboeuf-Yde, and Manniche 2003). The financial and economic burden of back pain, due to direct costs and work absence, is substantial (Lambeek et al. 2011; Manchikanti et al. 2014). Cost-effective interventions, focussing on prevention of LBP and return-to-work management, are essential to decrease its incidence and its burden on society.

There is strong epidemiological evidence that physical demands of work, such as manual materials handling and lifting, are associated with increased reports of back symptoms (Coenen et al. 2014; Heneweer et al. 2011; Griffith et al. 2012). Therefore, researchers and clinicians have focussed on reducing work-related risk factors for LBP by implementing interventions to decrease mechanical low-back load at work. According to the guidelines of the National Institute for Occupational Safety and Health (NIOSH), besides the mechanical load, physiological strain needs to be taken into account to guarantee safe manual material handling. High physiological strain can result in systemic or local fatigue (Waters et al. 1993), leading to an increased risk of lifting-related LBP. Janssens et al. (2010) found that systemic fatigue causes impaired coordination, potentially leading to an increased risk of low back injury. Furthermore, Wu and Wang (2002) have shown that there is a negative relationship between maximum acceptable work time and physical workload, measured in terms of aerobic strain. They

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recommend an upper limit of 34% $\dot{V}O_2$max (maximum rate of oxygen consumption (l/kg/min)) for dynamic work lasting 8 h. This suggests that high metabolic loads, as one component of physiological strain, should be avoided at work to prevent fatigue and low back pain injury.

Recently, body worn assistive devices, also called exoskeletons, have been introduced in work environments to reduce risk factors for LBP (de Looze et al. 2016). These devices physically support the user when performing tasks that involve high back loads. Several studies found reduced low-back mechanical loading during lifting, bending and static holding tasks when using assistive devices that passively support the user's trunk against gravity (Abdoli-E and Stevenson 2008; Graham, Agnew, and Stevenson 2009; Ulrey and Fathallah 2013; Wehner, Rempel, and Kazerooni 2009). By reducing internal moments, and hence muscle activity around the low back, or by allowing different movement strategies, which would otherwise put too much load on the low back without exoskeleton, these exoskeletons might also reduce metabolic load. Consequently, these exoskeletons might reduce fatigue and as such reduce this risk factor for injury.

A few studies have evaluated the effect of a body-worn lifting device on metabolic load during lifting. Whitfield et al. (2014) found that an on-body personal lift augmentation device (PLAD) reduces musculoskeletal effort but does not affect oxygen consumption during a continuous lifting task. Additionally, no change in lifting technique or movement strategy was found. Whitfield et al. (2014) suggested that this may be because some muscles got assisted by the device, while other muscle groups had to work harder. In the study of Sadler et al. (2011), greater ankle and hip flexion and less lumbar and thoracic flexion were found when wearing the PLAD system, indicating a change of lifting technique from a stoop lift to a 'semi-squat' technique. This change in technique could coincide with an increase in metabolic costs. Squat lifting has been found to involve higher metabolic costs than stoop lifting (Garg and Herrin 2007; Welbergen et al. 1991) due to higher muscle activity (Hagen et al. 1993) to make the body move through a larger range of motion, requiring more mechanical work. The increased metabolic costs associated with this change in technique might offset the potential reduction in metabolic costs from the unloading effect of the exoskeleton on the back muscles. This could account for the observed lack of change in metabolic costs while using the PLAD system. However, so far, the metabolic benefits (or costs) of only one passive trunk exoskeleton have been tested and thus, results cannot be generalised to other lifting devices.

Although potential positive effects of exoskeletons on mechanical and metabolic load are expected for specific load handling tasks, potential side effects of these devices on other tasks that need to be performed during working should not be ignored. Since workplaces nowadays are more versatile due to job rotation and automation (Dempsey 2002), a high variety of tasks beyond manual material handling and lifting is observed in work environments. In a previous study (Baltrusch et al. 2018), it has been established that tasks that involve large ROM (range of motion) of trunk or hip flexion, including walking, can be hindered by a trunk exoskeleton and are perceived as more difficult to perform with an exoskeleton. Participants also used a slower speed when walking with the exoskeleton. Thus, a device that supports the low back during lifting might require more muscle activity and hence increase metabolic costs during tasks such as walking.

The purpose of this study was to assess whether wearing a passive trunk exoskeleton affects the metabolic costs of repetitive lifting and walking. In addition, we explored which underlying changes in movement strategy and muscle activation patterns could explain these potential effects. It was hypothesised that wearing an exoskeleton during lifting reduces metabolic costs through a decrease in trunk muscle activity and/or change in lifting technique. In contrast, it was hypothesised that wearing the exoskeleton during walking increases metabolic costs. Assuming that people normally adopt an optimal step length that minimises energy cost (Bertram 2005), being forced to adapt step length due to the restriction by the device would likely increase energy costs.

2. Methods

2.1. Passive exoskeleton

In this study, the passive trunk exoskeleton 'Laevo' (Intespring, Delft, The Netherlands) was tested (Figure 1). This device is commercially available and in use at different work sites in various companies. It consists of four components: a pad at the anterior side of the chest, leg pads at the anterior side of the thighs, a pelvis belt to keep the device in a fixed position relative to the pelvis, and a smart joint with spring-like characteristics. The chest and thigh components are connected through semi-rigid bars running over this joint, which generates a supporting extension moment at the level of the low back when bending forward. To allow trunk rotation, the chest pad can rotate in the

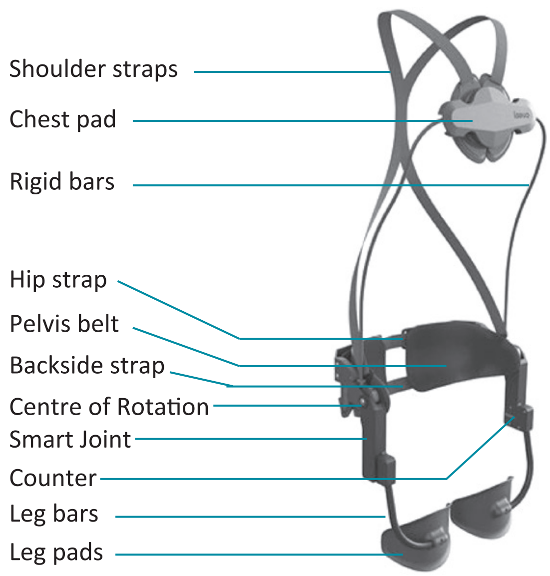

Figure 1. Laevo (Intespring, Delft, Netherlands). The user's trochanter major is to be aligned with the hip centre of rotation of the device. For personal length adjustment and for keeping the device in place, straps are provided. The chest pad facilitates walking by rotating in the frontal plane. Allowance of hip flexion/extension is regulated by the smart joint. The counter measures the number of hip flexions while wearing the device.

frontal plane of the trunk. Two types of the Laevo, with different angle-moment characteristics, were used for the lifting tasks in this study. According to the manufacturer, the high-cam Laevo predominantly supports the user at bending angles from 0 to 20 degrees, and the low-cam Laevo predominantly supports the user at bending angles >20 degrees.

2.2. Participants

Thirteen men with no prior history of low back pain participated in this study. Their mean age, height and body mass, here presented in mean (sd), were: 28.9 years (4.4), 1.80 m (0.04) m and 76.9 kg (12.0), respectively. Prior to the start of the measurement, participants received an information letter and signed informed consent. Approval for the experiment was given by the medical ethical committee of VU medical centre (VUmc, Amsterdam, The Netherlands, NL57404.029.16).

2.3. Instrumentation and data collection

2.3.1. Metabolic costs

The metabolic costs were determined through indirect calorimetry using a breathing gas analysis system (Cosmed srl, Quark CPET, Rome, Italy), measuring the rate of oxygen consumption and carbon dioxide production.

2.3.2. EMG monitoring

Muscle activity was recorded using surface Electromyography (Cometa Srl, Milan, Italy, Bandwith: 100 Hz, Input impedance: 20MOhm, Sampling Frequency: 2000 Hz). Surface electrodes were placed at 7 bilateral sites on the skin after abrasion and cleaning with alcohol (Ag-AgCl electrodes†; interelectrode distance, 20 mm). The recording sites were: m. longissimus thoracis (LT) at the T9 level (4 cm lateral), m. iliocostalis lumborum (IL) at the L2 level (6 cm lateral), m. longissimus lumborum (LL) at the L3 level (3 cm lateral), m. external oblique muscles (EO) about 15 cm above the SIAS, m. rectus abdominis (RA), 3 cm lateral from the umbilicus, m. vastus medialis (VM) and m. biceps femoris (BF).

2.3.3. Kinematic data

To analyse movement patterns during lifting and walking, we recorded 3D kinematics with an optoelectronic motion capture system (Optotrak, Northern Digital Inc., Waterloo ON, Canada) at a sample rate of 50 Hz. In the lifting trials, segment kinematics were measured using a dynamic three-dimensional linked segment model (Kingma et al. 2010). Cluster markers were attached to lower and upper leg, pelvis, trunk, upper and lower arm, head and box. Due to the fact that participants performed symmetric lifting, we only recorded kinematics from the right side of the body. To define the local segmental coordinate systems, anatomical landmarks were located through palpation and were related to the respective cluster markers using separate measurements (cf. Kingma et al. 2010). In the walking trial, only stride parameters were recorded using single heel markers attached to both shoes.

2.4. Testing procedure

Before starting the experimental trials, resting metabolic rate was measured with the participant sitting in a chair for 5 min, followed by fitting and adjusting the two trunk exoskeletons to the participant. Subsequently, participants were instrumented with EMG equipment and maximum voluntary isometric contractions (MVCs) were performed against resistance to normalise EMG data collected during subsequent trials.

Figure 2. Oxygen consumption, muscle activity and kinematics were measured during repetitive lifting (left) and walking (right).

The protocol was split into two parts: (1) Walking and (2) Repetitive Lifting (Figure 2). First, the walking tasks were performed, followed by the lifting experiment. Both protocols were performed on the same day.

In the walking protocol, a motorised treadmill (Forcelink, Culemborg, The Netherlands, dimensions: 3X5m) was used. To find preferred walking speed with and without the exoskeleton, a single marker was attached to the trunk of the participant. Subsequently, the participant walked on the treadmill in the 'self-paced mode', in which the treadmill adapts its speed to the speed of the participant based on tracking of the participant's position on the treadmill (Sloot et al. 2014). As soon as the treadmill reached a steady state, the walking speed was noted. After that, the oxygen consumption mask was fitted to the participant and single markers were attached to the shoes. The participant then performed two 5-min trials: normal walking without the exoskeleton (Control Condition) and walking with the low-cam Exoskeleton. These trials were performed at two different walking speeds: Preferred Walking Speed without the exoskeleton (PWS) and Preferred Walking Speed with the exoskeleton (PWSX), resulting in four different walking trials. The order of these trials was randomised for each participant. Note that only the low-cam Exoskeleton was used in the walking experiments since this type of exoskeleton should, in theory, provide the least hindrance during walking, based on the fact that it provides support at high angles. A walking period of 5 min was chosen to achieve steady-state walking. Three minutes of rest were provided between conditions to prevent fatigue.

Before starting the lifting protocol, participants had a break of 30 min to prevent fatigue. After attaching the cluster marker and fitting the oxygen mask again, participants performed the repetitive lifting task. They were required to lift and lower a 10-kg box ($0.39 \times 0.37 \times 0.11$ m, with 2.5 cm diameter handles) at a rate of 6 lifts per minute. Each lift consisted of picking up the box, assuming an upright posture, putting down the box, and assuming an upright posture again. The participants were instructed to choose their own lifting technique. In that way, we took into account potential changes in lifting technique that will occur when people use this exoskeleton in practice. The lifting rate was imposed by a metronome that sounded in each upright position. Movement speed was left free, as participants were allowed to pause the movement when standing upright. Hence, the movement was performed at a natural speed. Each lifting trial lasted for 5 min and was performed in three conditions: lifting without the exoskeleton (Control Condition), lifting with the high-cam exoskeleton and lifting with the low-cam exoskeleton. These lifting trials were conducted from two different heights: ankle height and knee height. The order of the trials was block randomised per exoskeleton condition. Approximately 30 s of rest was given between trials.

2.5. Data processing

Metabolic energy expenditure (J/min) was calculated from $\dot{V}O_2$ (ml/s) and respiratory exchange ratio (RER; Garby and Astrup 1987). Flow rates were averaged over the final 2 min to ensure a steady state condition and were normalised to body mass. Net metabolic energy expenditure was calculated by subtracting resting metabolic rate from the total metabolic rate during walking and lifting. Net metabolic cost for walking was obtained by normalising net metabolic energy expenditure to walking speed and was expressed in J/kg/m. For lifting, net metabolic cost was not normalised to speed and was expressed in J/kg/s.

EMG signals were high-pass filtered with a 2nd order Butterworth filter (cut-off frequency 20 Hz, bidirectional) to remove movement artefacts. Additionally, a 4th order Butterworth band stop filter (49–51 Hz, bidirectional) was applied to remove hum. After rectifying the data, the data were filtered with a 4th order low pass Butterworth filter (cut-off frequency 4 Hz) to create a linear envelope. Next, we normalised muscle activity for each muscle to the maximum of the linear envelope obtained in the MVC trials. EMG envelopes were then normalised to cycle time and averaged over cycles.

Kinematic data were filtered with a 2nd order low pass Butterworth filter with a cut-off frequency of 5 Hz. For the lifting trials, the instantaneous 3D knee, hip, trunk and lumbar joint angles were calculated (Kingma et al. 2010). We only analysed angles in the sagittal plane of the anatomic reference frames, hence: knee flexion/extension, hip flexion/extension, trunk inclination and lumbar flexion/extension. Knee peak angles were used to define a lifting cycle. One lifting cycle was defined as bending down, lifting the box, holding the box in upright position, lowering the box, and returning to upright position without the box again. Range of motion per joint was calculated and averaged over movement cycles, generating an average value for each condition per participant. The centres of mass of all segments were used to calculate the total body centre of mass. By taking the mean of the vertical distance travelled by the body centre of mass over each separate lifting cycle, we arrived at the average range of motion of the body centre of mass during a lifting cycle.

For the walking trials, heel strikes for each cycle were determined, using the data of the two heel markers. Left and right heel strikes were defined as the instants of local maxima in the horizontal position of the respective heel marker in the sagittal plane. Heel strikes were plotted and visually checked for detection errors. The heel strikes were used to calculate average stride times and stride lengths per trial. Stride times were calculated as the time differences between subsequent left heel strikes. Stride lengths were calculated as the distances between subsequent left heel strikes, correcting for speed of the treadmill and duration of the stride. Heel strikes were also used to define gait cycles for the EMG data, which were normalised to cycle time and averaged over cycles.

2.6. Statistics

To test for the effect of exoskeleton use on selected dependent variables, we conducted one-way repeated-measures ANOVAs. We a-priori decided not to test for the main effect of lifting height or walking speed and their interaction with exoskeleton. Thus, we conducted two separate ANOVAs for each of the lifting heights (knee and ankle height) and two separate ANOVAs for each walking speed (PWS and PWSX). Hence, for the lifting task, these ANOVAs included 1 factor (exoskeleton) with 3 levels (high-cam Laevo, low-cam Laevo, control condition without). For walking the ANOVAs included one factor (exoskeleton) with 2 levels (low-cam Laevo, control condition without). In case of a significant effect of exoskeleton in the lifting experiment, Bonferroni post-hoc tests were conducted to determine differences between exoskeleton conditions. Alpha of 0.05 was used as the critical level of significance. All statistical analyses were performed using SPSS for Windows (IBM, SPSS Statistics 23.0, USA). To test for statistically significant differences in muscle activity, we used one-dimensional statistical parametric mapping (SPM1D) (Pataky et al. 2013) to perform an SPM1D repeated measure ANOVA between the exoskeleton conditions high-cam Laevo, low-cam Laevo and control for lifting and low-cam Laevo and control for walking. This analysis uses random field theory to make statistical inferences on the time intervals over which the independent variable (exoskeleton use) has a significant effect on muscle activity. In view of some loss of individual data for different outcome measures, the number of participants (*N*) included in the statistical calculation is reported for each outcome.

3. Results

Metabolic costs and underlying changes in movement strategies and muscle activity are reported first for lifting, followed by walking.

3.1. Lifting

3.1.1. Metabolic costs

A main effect of exoskeleton use was found on metabolic costs during lifting for both, knee height ($p = .046$) and ankle height ($p = .047$, Figure 3). For lifting from knee height, post-hoc testing revealed a significant reduction in metabolic costs of 17% between Control condition and High-cam Exo condition (mean (sd): 3.09W/kg (0.92) vs. 2.56W/kg (0.52); $p = .012$). For lifting from ankle height we found 16% decrease in metabolic costs between Control condition and High-cam Exo condition (mean (sd): 5.06W/kg (1.11) vs. 4.27W/kg (0.60); $p = .012$). The 7% reduction in metabolic cost between Low-cam Exo and Control condition when lifting from knee height and 8% reduction when lifting from ankle height did not reach significance.

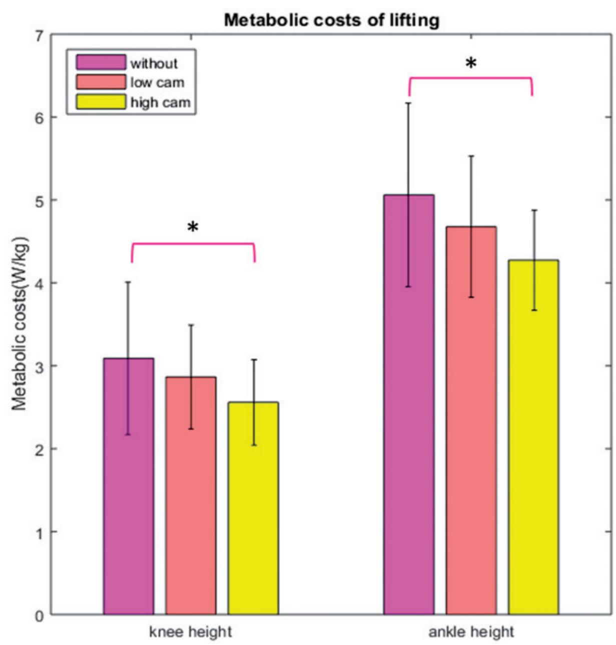

Figure 3. Left: Metabolic costs of lifting from knee and ankle height. Values are normalised for bodyweight. $N = 11$. Error bars indicate standard deviations. *Significant change in metabolic costs between control condition (without) and exoskeleton condition (low cam/high cam).

3.1.2. Kinematics

With regard to movement strategies in lifting, the range of motion (ROM) in the joints analysed did not show a main effect of exoskeleton use (Figure 4). Still, a tendency to smaller range of motion in all joints was observed in the exoskeleton conditions when lifting from ankle height.

The average range of motion of the centre of mass (COM) did not show a significant difference between the exoskeleton conditions when lifting from knee height. However, in line with joint ROM, when lifting from ankle height, the range of motion of the COM tended to be lower when wearing the exoskeleton, compared to control condition (Figure 5a). This approached significance ($p = .056$). Figure 5b,c shows the lowest and highest position of the COM, averaged over participants. Again, no significant differences

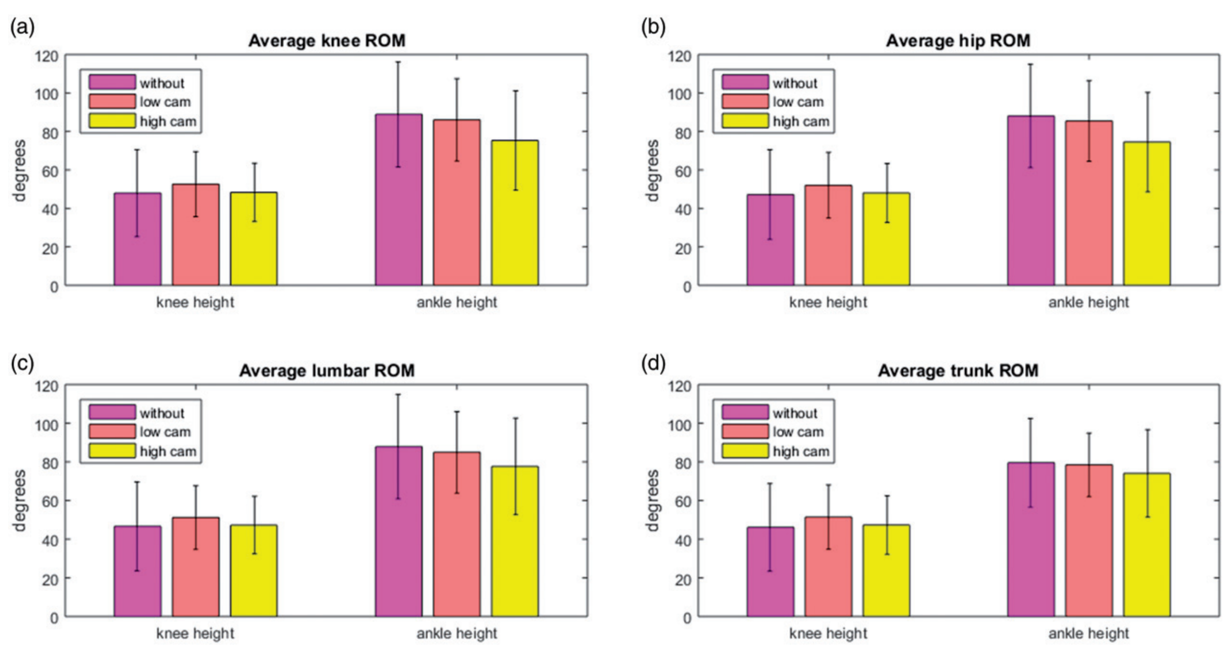

Figure 4. Range of motion in the knee joint (a), hip joint (b), LS51 joint (c) and trunk (d) when lifting from knee and ankle height, averaged over all participants. $N = 11$. Error bars indicate standard deviations.

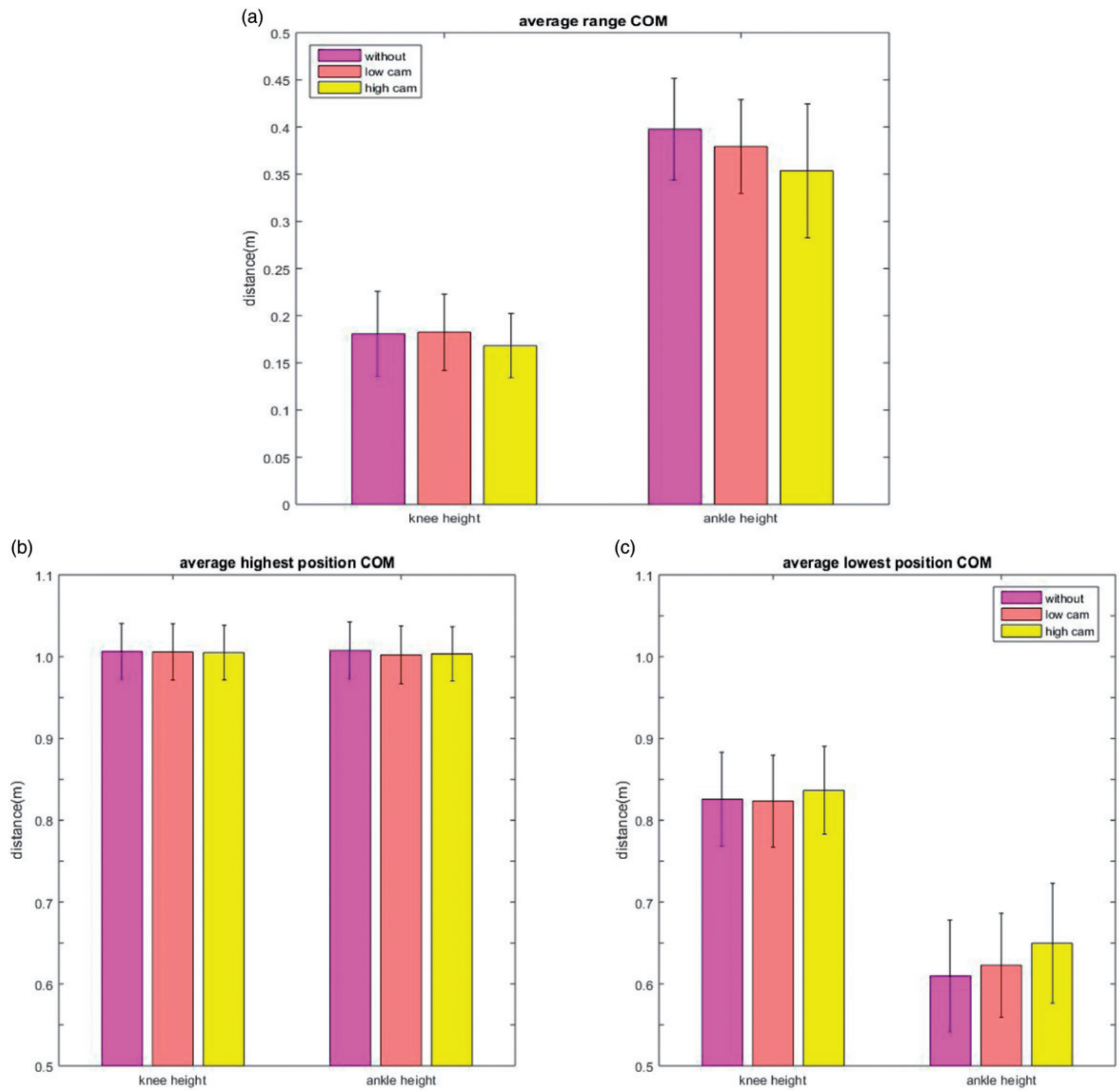

Figure 5. The amplitude of the centre of mass when lifting from knee and ankle height, averaged over all participants (a). Highest position of the centre of mass when lifting from knee and ankle height, averaged over all participants (b). Lowest position of the centre of mass when lifting from knee and ankle height, averaged over all participants (c). $N = 11$. Error bars indicate standard deviations.

between the exoskeleton conditions were found, but the figure demonstrates that the potential difference in range of motion of the COM mostly resulted from a decrease in downward movement when wearing the exoskeleton and not from a lack of extending upwards.

3.1.3. Muscle activity

The muscle activity of the trunk muscles over a lifting cycle is shown in Figure 6a (lifting from knee height) and Figure 6b (lifting from ankle height). Although on average peak muscle activity of the back muscles seemed to be lower when wearing the exoskeleton, especially when lifting from ankle height, this difference was not significant in any period of the lifting cycle. In contrast, abdominal muscles did show significant main effects of exoskeleton for both lifting heights. The muscle activity of m. rectus abdominus showed a small but significant increase when lifting with the exoskeleton from knee height ($p = .050$). M. external oblique significantly increased in activity when lifting with the exoskeleton from ankle height ($p = .042$). Post-hoc analysis suggested that this latter main effect derives from a difference in activity between low cam and control condition although it did not reach significance. We did not find any effects of the exoskeleton on muscle activity in the upper legs.

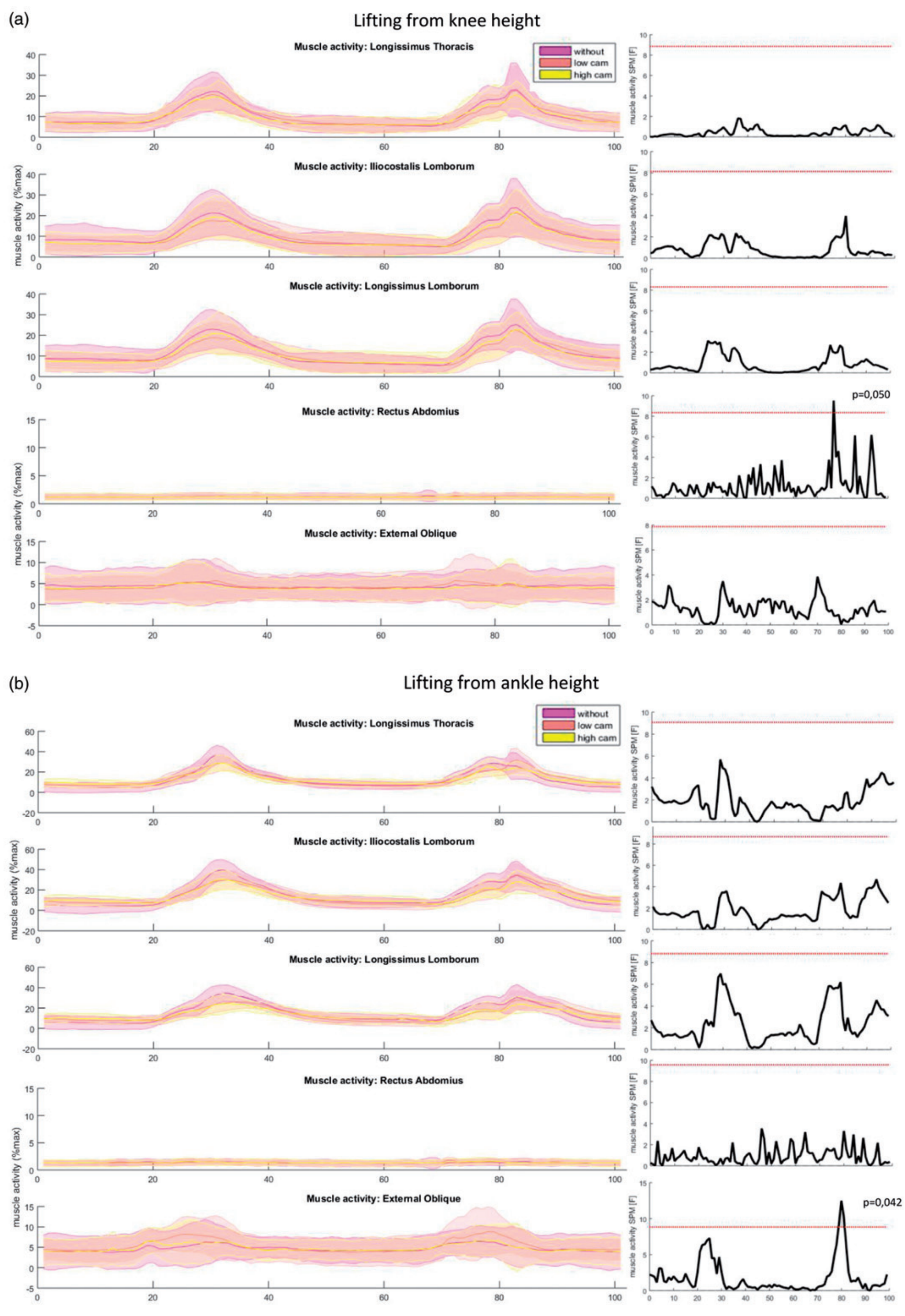

Figure 6. (a) Left: Averaged time series of muscle activity per cycle for each condition, averaged over participants when lifting from knee height. The shaded errors represent the standard deviation. $N = 11$; Right: The relative one dimensional repeated measures ANOVA (SPM1D) of muscle activity of the control condition compared to the exoskeleton condition. The horizontal axis displays normalised cycle time. The vertical axis displays the one dimensional F-statistic. A significant effect is present at instances where the black line is above the horizontal dotted red line. (b) Left: Averaged time series of muscle activity per cycle for each condition, averaged over participants when lifting from ankle height. The shaded errors represent the standard deviation. $N = 11$; Right: The relative one dimensional repeated measures ANOVA (SPM1D) of muscle activity of the control condition compared to the exoskeleton condition. The horizontal axis displays normalised cycle time. The vertical axis displays the one-dimensional F-statistic. A significant effect is present at instances where the black line is above the horizontal dotted red line.

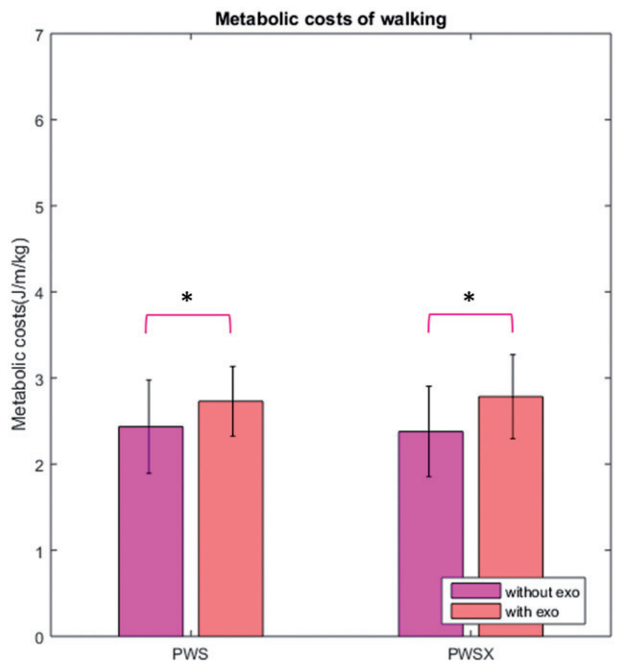

Figure 7. Metabolic costs of walking in preferred walking speed without exoskeleton (PWS) and preferred walking speed with exoskeleton (PWSX). Values are normalised for body-weight and walking speed. $N = 13$; Error bars indicate standard deviations. *Significant change in metabolic costs between control condition (without) and exoskeleton condition (with exo/low cam).

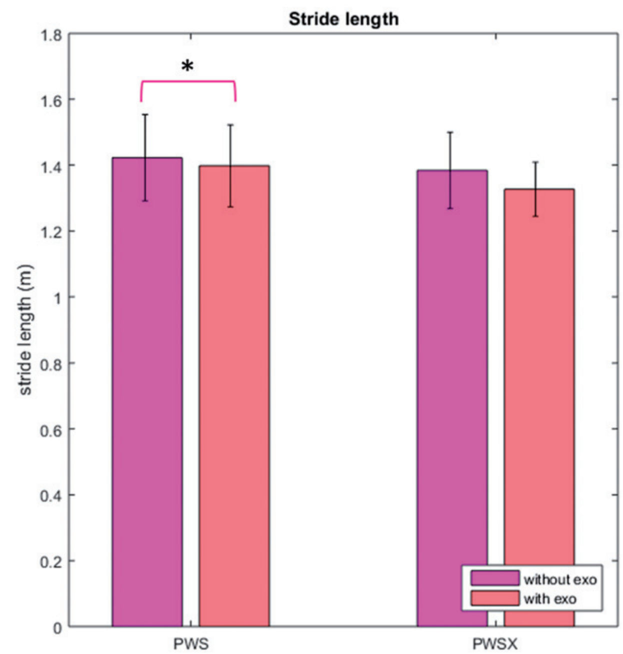

Figure 8. Stride length when walking with and without exoskeleton in the preferred walking speed without exo (PWS) and in the preferred speed with exo (PWSX). $N = 10$; Error bars indicate standard deviations. *Significant change in stride length between control condition (without) and exoskeleton condition (with exo/low cam).

3.2. Walking

3.2.1. Metabolic costs

For walking, metabolic cost increased by 12% and 17% when wearing the exoskeleton in speed conditions PWS and PWSX, respectively (Figure 7). For both speed conditions, this effect was significant (PWS: $p = .002$ and PWSX: $p = .002$).

3.2.2. Kinematics

The preferred walking speed without exoskeleton (PWS) and the preferred walking speed with exoskeleton (PWSX) were slightly, but significantly different. Participants preferred to walk faster without the exoskeleton than with the exoskeleton (mean (sd): 1.27 m/s (0.16) vs. 1.22 m/s (0.14); $p = .05$). A reduction in stride length was found when walking with exoskeleton compared to without (Figure 8). This effect was, however, only significant at the preferred walking speed determined without exoskeleton (mean (sd): 1.42 m (0.13) vs. 1.40 m (0.13); $p = .013$).

3.2.3. Muscle activity

The muscle activity of the trunk muscles over a stride is shown in Figure 9. The muscle activity of the back muscles did not show significant differences between the exoskeleton conditions at either of the walking speeds. With regard to the abdominal muscle activity, a significant increase in muscle activity of the m. rectus abdominus was found when walking with the exoskeleton at PWSX, at three different time instances during the gait cycle ($p = .036$, .049 and .050). The muscle activity of m. external oblique significantly increased in the ipsilateral initial swing when walking with the exoskeleton at PWS ($p = .041$), and in the ipsilateral mid stance when walking with the exoskeleton at PWSX ($p = .026$). Muscle activity in the legs did not show any significant differences between conditions.

4. Discussion

The main goal of this study was to evaluate the effect of a passive trunk exoskeleton on metabolic load during lifting and walking. As hypothesised, we found a decrease in metabolic costs when wearing the exoskeleton during lifting and an increase in metabolic costs when wearing the exoskeleton during walking.

4.1. Lifting

Our results reveal that wearing the exoskeleton during lifting decreases metabolic costs. However, we only

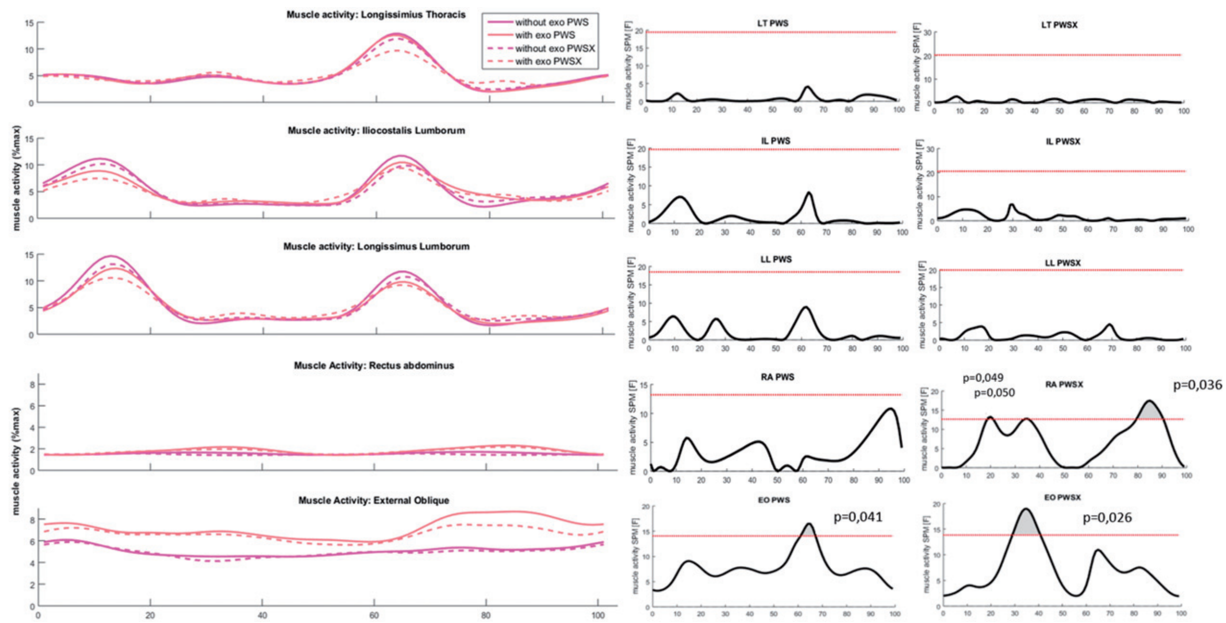

Figure 9. Left: Averaged time series of muscle activity per stride for each condition, averaged over participants when walking with and without exoskeleton in the preferred walking speed without exo (PWS) and in the preferred walking speed with exo (PWSX) $N = 9$; Right: One dimensional repeated measures ANOVA (SPM1D) of muscle activity of the control condition compared to the exoskeleton condition. The horizontal axis displays the normalised stride cycle. The vertical axis displays the one-dimensional F-statistic. A significant effect is present at instances where the black line is above the dotted horizontal red line.

found a significant difference with respect to the control condition when using the high-cam exoskeleton, which yielded reduced metabolic costs of up to 17%. Wearing the low-cam exoskeleton metabolic costs showed a modest reduction of up to 8%, which did not reach significance.

Our results are not in line with Whitfield et al. (2014) who did not find a change in oxygen consumption when lifting a mass of 9 kg for 15 min with a personal lift assistive device (PLAD). A possible explanation can be found in Sadler et al. (2011) who tested the effect of the same device (PLAD) on lifting technique and found greater hip flexion and less lumbar flexion. This suggests a change from a stoop towards a more squat-like technique when using the device. Since this squat technique requires more metabolic energy than a stoop technique, due to higher muscle activity in the legs, these findings might explain the lack of effect on metabolic costs when wearing the PLAD system. This is in line with Whitfield et al. (2014) who suggested that the lack of effect could be caused by the fact that some muscles may have been assisted while other muscles had to work harder when wearing the device.

In contrast to the results found for the PLAD system, in the present study, participants changed their lifting technique to a stoop-like technique when using the exoskeleton, reducing their COM movement amplitude. The reduced downward motion of the COM requires less mechanical work to be generated against gravity. To compensate for the decrease in COM movement, participants possibly extended their arms more at the lowest point of the lift to pick up the box from the designated height. This change of movement strategy may contribute to the significant decrease in metabolic consumption when lifting from ankle height that we found in our study. These differences in response between studies may derive from the different designs of the exoskeleton. While the design of the PLAD provides a connection of the pelvis part to the legs through elastic latex bands, running to the lower legs, the Laevo consists of leg pads on the anterior sides of the thighs. As during lifting the exoskeleton transfers forces from the low back to these leg pads, resistance occurs when squatting, favouring less pronounced squatting with Laevo.

A tendency to a decrease in back muscle activity was observed when wearing the Laevo exoskeleton, especially when lifting from ankle height (Figure 7b). This indicates that the required torque of the trunk extensor muscles is partly supported by the torque generated by the Laevo although this effect was smaller than expected. In addition, we found a small but significant increase in the activation of abdominal muscles when lifting with the exoskeleton, which was especially pronounced in the low-cam condition. This indicates that participants increased abdominal activity to overcome the resistance of the exoskeleton during trunk flexion. It is arguable whether this change in muscle activity influenced metabolic costs since the increase is rather small. However, the effect on abdominal activation may partly account for reduced beneficial effects of the low-cam exoskeleton on metabolic costs of lifting compared to the high-cam exoskeleton. Bosch et al. (2016) evaluated the effect of the Laevo on muscle activity during an assembly task in a forward bent position. They found that muscle activity in the lower and upper back decreased by 38% and 44%, respectively when the participants used the exoskeleton. Abdominal muscle activity did not change. The participants in their study had to perform a pick and place task, requiring work in a static position with the trunk bent forward to 40 degrees flexion. Due to the fact that their participants performed a static task, continuous support was provided by the Laevo exoskeleton, which was not the case in the present study due to the dynamic behaviour of the lifting task. This explains the larger effects on back muscle activity found by Bosch et al. (2016) compared to the present study. Finally, Bosch et al. (2016) and Whitfield et al. (2014) found decreased activity in the M. biceps femoris when wearing an assistive device. We did not find any changes in the leg muscles, which may be due to differences between tasks investigated and techniques to perform these tasks.

Since lifting requires trunk inclination of more than 20 degrees, we expected a bigger effect of the low-cam exoskeleton, which is supposed to support the user at bending angles >20 degrees, compared to the high-cam exoskeleton, which is supposed to support the user at bending angles from 0 to 20 degrees. However, the low-cam exoskeleton showed smaller and non-significant effects on metabolic costs. We, therefore, assessed the torque-angle characteristic of the device in a 'post-hoc' measurement, to test whether it matches the description of the manufacturer. Using a force transducer, we measured chest pad forces when a participant was performing trunk bending motions through the full range of motion of the exoskeleton. Additionally, we assessed the angle of the exoskeleton hip joint using Optotrak motion capture markers (Koopman et al. 2019). The results of this measurement are shown in Figure 10.

The maximal support of the high-cam device is reached at a Laevo angle of about 35 degrees. This is within the range of joint angle that was reached during lifting. This in contrast to the low-cam exoskeleton that provides a maximal support at a joint angle >100 degrees, which was never reached by the participants during lifting. The steep incline of the torque in the end range of motion of both types is due to a hard stop that is provided by the device. These findings reveal that the high-cam Laevo provided the user with more support over the range of motion relevant to our task than the low-cam Laevo at both lifting heights, and explains the significant reduction in metabolic costs for the high-cam exoskeleton as opposed to the low-cam exoskeleton. Another important characteristic of both Laevo exoskeletons is the hysteresis effect. The moments provided by the exoskeleton are higher for the flexion phase than for the extension phase, thus support and stored mechanical energy get lost in the system during the movement. This explains the larger effects on back muscle activity found in the static task by Bosch et al. (2016) compared to dynamic task in the present study.

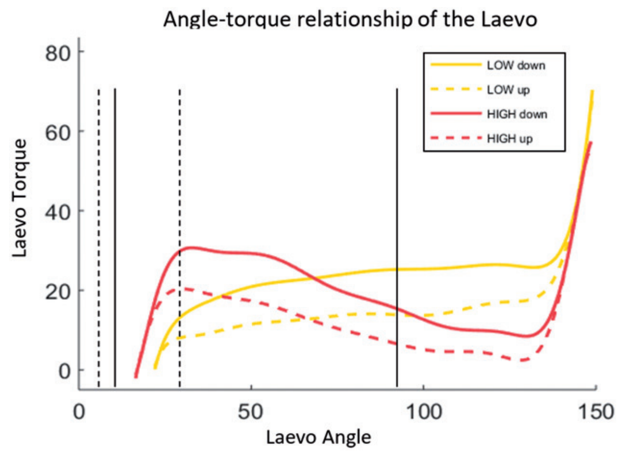

Figure 10. The angle-torque relationship of the high- and low-cam exoskeleton. The vertical black lines represent the operating range of the device during lifting from knee height (dashed line) and ankle height (solid line) (adapted from Koopman et al. 2019).

The lack of statistical significance in outcome parameters of muscle activation and movement strategy, which underlie the observed reduction in energy cost, shows that individual participants responded differently to the exoskeleton when lifting. Individual participants changed their lifting strategy from squat to stoop to different extents. This resulted in inter-individual differences in COM movement changes, and different changes in muscle activation patterns to arrive at the consistently reduced metabolic costs. Statistical power of this study, however, was not sufficient to perform subgroup analysis. Determining which of the underlying factors accounts most for the observed reduction of metabolic costs requires further research.

4.2. Walking

Wearing the exoskeleton during walking increased metabolic costs by 12% at PWS and 17% at PWSX, confirming our second hypothesis. The significantly slower preferred walking speed with the exoskeleton compared to walking without the exoskeleton indicates that the exoskeleton hinders walking. The changes in movement strategy underline that. Participants shortened their steps when walking with the exoskeleton at the faster speed, suggesting that they probably had to cope with the resistance of the device against hip flexion. This is in line with a previous study (Baltrusch et al. 2018), which tested the effect of the same exoskeleton on functional performance. The results yield increased perceived difficulty of tasks that involve hip flexion, underlining the importance of the possibility to disengage the device to allow unrestricted hip flexion in walking and similar tasks.

While the exoskeleton did not have any effect on back muscle activity during walking, effects on abdominal muscle activity were found. When walking with the exoskeleton the activity of m. rectus abdominus and m. external oblique increased significantly. This is another indication of the impeding effect of the exoskeleton during walking. Participants increased their abdominal muscle activity to cope with the resistance against hip flexion. When flexing the hip, the abdominal muscles stabilise the pelvis in the sagittal plane and prevent it from anterior tilting by the downward pull of the hip flexor muscles (Neumann 2010; Hu et al. 2012). In case of restricted hip flexion, participants probably also increased activity of their hip flexors. However, hip flexor activity is difficult to assess with surface EMG.

In summary, walking with a passive trunk exoskeleton increases metabolic cost. Indicators for restricted hip flexion were found in both movement strategy and muscle activity. However, it is unclear whether the increase in metabolic costs can be solely explained by these indicators.

4.3. Limitations

Our protocol was performed in a laboratory setting and limited to a lifting time of 5 min, unlike real-life work settings, where lifting tasks are often much more variable in terms of technique and frequency. Thus, the outcome of this study cannot be directly generalised to a normal working environment, further studies are needed to assess the effect of a passive trunk exoskeleton on metabolic costs during a whole working day. Furthermore, due to data loss in the kinematic and EMG analysis, the statistical power for these parameters was lower than for our main outcome, metabolic cost, which explains why we found a significant effect in our main outcome but only trends in the underlying mechanisms. Finally, due to the different designs of the various exoskeletons that are currently assessed in research, we cannot generalise our outcome to other assistive devices since effects are dependent on the design of the exoskeleton.

5. Relevance and conclusion

We have shown that wearing a passive trunk exoskeleton decreases metabolic costs during lifting and increases metabolic costs during walking. The remaining question is whether the observed effects have a meaningful effect on fatigue in daily practice. People tend to operate at ~36% of their maximal aerobic capacity to avoid fatigue (Astrand et al. 2003). According to Wu and Wang (2002), employees should not exceed 34% of their aerobic capacity when working shifts of 8 h. This is in line with other research (Michael et al. 1961; Bink 1962; Lehmann 1983; Astrand et al. 2003), which recommends that the level of oxygen consumption should not exceed 33% of $\dot{V}O_2max$ for working in shifts that last between 2 and 8 h. To understand the relevance of our results, we can express our observed effects in similar terms of relative aerobic load. Assuming that our participants walked at 36% of their aerobic capacity in our self-selected walking speed trials (confirm the finding of Astrand et al. 2003), we can estimate that lifting from knee and ankle height without the exoskeleton required ~36% and ~51%, respectively. Wearing the high-cam exoskeleton reduced these values to ~33% and ~47% oxygen consumption. Although the reduction in relative load by 3–4% appears small, it may be relevant for the working population, considering that aerobic load of repetitive lifting in this study is around or exceeding the recommended maximal aerobic load indicated in the literature. Thus, reducing the net metabolic consumption with the use of an exoskeleton is a relevant possibility to enhance safe work without undue fatigue. Certainly, future studies are needed to prove this statement.

Our findings suggest that exoskeletons are of benefit for lifting by decreasing physiological strain. Work-related low-back pain, in particular, might be preventable when wearing an exoskeleton, due to a lower risk of getting fatigued. Data on underlying changes in muscle activity and movement strategies provided insights for further optimisation of exoskeleton design from the perspective of metabolic costs. Future studies are needed to corroborate underlying mechanisms and design optimizations.

Acknowledgements

The authors would like to acknowledge the support of Laevo for unconditionally providing the exoskeleton for this research.

Disclosure statement

No potential conflict of interest was reported by the authors.

Funding

The work presented in this paper was supported by the European Union's Horizon 2020 research and innovation programme under grant agreement No 687662 – SPEXOR.

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